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6333546001

4. Title of the invention

IMAGING

5. Name of your agent (if you have one)

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Imaging

5 This invention relates to imaging and in particular to methods and systems for image enhancement.

10 Imaging is carried out by transfer from the object domain into the image domain, but owing to limiting factors such as the finite size of the energy source, detector size, sampling frequency, display density, software filter function, and possibly partial-volume effects experienced with some imagers, an infinitely fine delta function in the object domain cannot be faithfully reproduced in the image domain. Instead, a smeared-out image, or point-spread function (PSF), is observed. Similarly, an infinitely sharp edge-response function (ERF) in the object domain becomes a smeared-out ERF in the image domain. When two image ERFs are close to each other, the smeared-out effects run into one another. The deterioration of the spatial resolution is often monitored by the percentage modulation transfer which is given by the ratio, expressed in percentage, of the amplitude of the modulation in the image domain to that of the object domain.

25 The smearing effect becomes more intense as the adjacent ERFs of discontinuities or contrast profiles get closer to each other (or as the spatial frequency of the modulation becomes higher); and this also causes the loss of profile heights. The modulation transfer function (MTF) is the function of the 'normalised' percentage modulation transfer over the whole range of spatial frequency until the cut-off frequency at which the amplitude of the image modulation transferred becomes zero, or, in practice, when it falls below the noise level and becomes undetectable.

The shape of the overall MTF over the full range of spatial frequency is similar to a roll-off sinc function, and the high contrast profile has a higher cut-off spatial frequency than a low contrast profile. Within
5 the working range before the cut-off spatial frequency, the inherent loss of the spatial resolution (that is, the part that is indicated by the smeared-out effect on the corner edge of the ERF) cannot be restored or partially restored even by re-scanning the image with an ultra high
10 resolution digital scanner system.

It is an object of the present invention to provide a method and system by which the above problem can be overcome at least partially.

15 According to one aspect of the present invention there is provided a method of image enhancement wherein a de-convolution process is applied to the image-domain results of an object-scan to re-construct therefrom the
20 respective point-spread function (PSF) effective in the object- to image-domain transfer at one or more object-discontinuities, and correlating this re-constructed function with the image-domain results of those one or more discontinuities for image-resolution enhancement as
25 to discontinuity-location in the image domain.

According to another of the aspects of the present
~~invention there is provided an image-enhancement system~~
30 comprising means for performing a de-convolution process on the image-domain results of an object-scan to re-construct therefrom the respective point-spread function (PSF) effective in the object- to image-domain transfer at one or more object-discontinuities, and means to
35 correlate this re-constructed function with the image-domain results of those one or more discontinuities for image-resolution enhancement as to discontinuity-location in the image domain.

A least-squares running filter may be used for the de-convolution process.

5 In the method and system of the present invention the optimum edge position corresponding to the discontinuity or sharp-edge position of the object ERF may be pin-pointed from the central point of the full-width half-maximum (FWHM) of the image PSF; the pin-pointing may be to sub-pixel accuracy for the image ERF. The original
10 discontinuity or sharp-edge feature of the ERF may then be restored within the image domain by removing the sub-pixel values from outside the optimum edge position to compensate for those within. It is to be noted that the sub-pixels become pixels, and that the enhancement is
15 equivalent to the performance of an extra high resolution image transfer system.

It may be shown mathematically that the image ERF of an infinitely-sharp edge-response function ERF can be
20 reproduced by convoluting the image PSF with the object ERF. This is illustrated in Figure 1 of the accompanying drawings. There is also illustrated in Figure 1 how the image ERF is used in conjunction with the image PSF to determine the optimum edge position from the full-width
25 half maximum FWHM. The optimum edge position corresponding to the sharp edge position of the object ERF may be pin-pointed to sub-pixel accuracy for the
image ERF from the central point of the FWHM.

30 For one-dimensional cases, the operation in accordance with the invention is relatively simple, as only either the X or the Y profile (that is to say, as indicated below, a line spread function (LSF)) is involved). But for the two-dimensional operations, both the X and Y
35 profiles, and if necessary, the XY diagonal profiles to eliminate any possible streakings in the image, may be

involved. In this case, a proper weighting scheme will be required to re-construct the image.

5 Once the original sharp-edge feature has been pin-pointed, that feature may be restored within the image domain by removing the sub-pixel values from outside the optimum edge position and utilising them for compensation of those within. This enhancement operation is illustrated in the form of the modulation display of
10 Figure 2 of the accompanying drawings.

The technique of the invention may partially recover the loss of the spatial resolution without the trade-off loss of other properties such as image noise. Furthermore,
15 the enhancement of the spatial resolution in an image, of the invention may lead to enlargement or magnification of a small area or particular region of interest (ROI) without blurring effects at the profile edge, for example as when the sub-pixels are displayed as pixels at the
20 ROI.

A feasibility study using a theoretical data set and the noisy low-contrast magnetic resonance (MR) images of pig brain and lamb kidney surrounded by water have been
25 successfully carried out. The results are illustrated in Figures 3 to 6 of the accompanying drawings, in which Figure 3 shows results for a theoretical data set,
30 Figures 4 and 5 the results achieved for pig brain, and Figure 6 for lamb kidney. The signal-to-noise figures of the re-created PSF (measured by the main peak height to that of the adjacent peaks) were over 8/1.

Figure 7 of the accompanying drawings is a flow chart illustrative by way of example of the image-enhancement
35 system.

The method and system of the invention are applicable, for example, in medical science, in engineering, in physical science and in the field of instrumentation generally. In medical science the invention may be
5 utilised in connection with magnetic resonance (MR) imaging, computer assisted tomography (CT), X-ray radiography, digital fluorography, ultra-sound imaging, nuclear medicine, proton emission tomography (PET) and other camera or imaging. In MR, X-ray, and CT it may be
10 used, for example, to pin-point the true boundaries of tumour growth or any other abnormal effects essential for the diagnosis, surgery, and radiation treatment. Moreover, the technique is particularly suitable for use in X-ray digital fluorography, in which small structures
15 under study are highlighted by injection of contrast liquids; the small structures may also be isolated from surrounding interfering effects by using an image subtraction technique. The present high-resolution enhancement technique may be used to improve the image
20 quality further.

Image subtraction coupled with the present resolution-enhancement technique may also be applied, for example, to MR images from two different scan-sequences. Also,
25 the magnification of an image ROI without blurring effects at the edges of the profile has many uses.

The inherent resolutions of X-ray radiography, ultra-sound imaging, nuclear medicine, and PET scanning
30 are relatively low, and some are used for real-time study. Only the individual still frame or hard-copy images may be re-processed.

In the context of engineering, physical science and the
35 field of instrumentation, the invention is applicable to one-dimensional imaging as used, for example, in regard to bar-code patterns, the spectrum of DNA analysis, iris

patterns of eyes (for example, for identification purposes in commercial banking), finger-print identification, and emission spectroscopy. The invention is also applicable to two-dimensional imaging, for example, in relation to images obtained by satellite or pattern recognition, or from a surveillance camera or during laboratory experimentation. As a general matter, the invention is applicable where there needs to be accurate determination of the edge position in an image versus the true object-edge position, for the purpose, for example, of measurement of the positional displacement between object and image, distortion correction and manufacturing control.

As an example of one of many applications of the present invention, a method and system that uses CT and MR imaging in conjunction with one another, will now be described.

With CT, signal magnitude is monitored by the attenuation of the polychromatic X-ray photon energy by all the atoms within the beam path. The linear attenuation coefficient, μ , is dependent on the density of the absorber and the atomic numbers of the atoms in the X-ray beam path, and contrast numbers relative to water on an absolute scale are derived. However, MR makes use of signals derived from the so-called 'T1-weighted' and 'T2-weighted' protons of quantum processes, and has contrast numbers that are relative in scale. The MR contrast numbers are accordingly uncorrelated with those for CT. This lack of correlation is confirmed by consideration of bone and water which are at opposite ends of the CT contrast (absolute) scale, but which are at the same end of the MR-image contrast (relative) scale owing to their common low proton-population.

The lack of correlation between the CT and MR signals acts against their use in combination for imaging purposes, but the present invention provides a method and system by which the advantages of each may be utilised to
5 improve image resolution and contrast information.

In the latter regard, CT provides a high spatial resolution but only in regard to view normal to the transverse slice. Resolution for all re-constructed
10 non-transverse planes is poor owing to the need to use elongate voxels to improve signal-to-noise ratio. Also, the partial-volume effect of using elongate voxels may give rise to detection errors at the thin edge of a contrast profile of a lesion. MR, on the other hand, can
15 give the same high degree of spatial resolution viewed in the normal direction to any image-slice plane, and can also provide isotropic resolution with cubic-voxel volume imaging.

20 To this end, multiple-slice transverse CT scans are collected across a section of the volume of interest in a patient or other subject. Corresponding multiple-slice transverse MR scans of the same volume are collected correspondingly. The slice thickness of the latter scans
25 may be one half, or smaller, of the thickness of the CT slices, and may be collected two-dimensionally or three-dimensionally. The patient is constrained throughout on a couch that provides landmarks with respect to a coordinate reference arrangement on the couch-top. This
30 is to ensure the reproducibility of anatomical positions and features to the first order accuracy for the corresponding CT and MR scans, and also for the final radiation treatment to be made.

35 The respective transverse planes of the CT and MR images are processed individually in the method and are matched with one another in a de-convoluted space for the CT and

MR images, as later explained. The two sets of de-convoluted maps are then merged together to a second order of accuracy in order that the CT numbers may be transferred over to replace the corresponding MR contrast numbers. Once this has been achieved, non-transverse (or oblique) planes can be obtained from the two-dimensional (2D) MR images or from the re-arrangement of the corresponding voxels of the three-dimensional (3D) volume images; where a 2D-MR is used, a further step of contrast transformation may be required.

In the method, CT multiple-slice transverse (Tr) scans are collected across a section of the volume of interest in a patient. Similarly, corresponding MR multiple-slice transverse scans of one half, or smaller, of the slice thickness of the CT slices are collected across the same volume of interest by 2D-MR sampling or 3D-MR sampling.

The patient is constrained from movement during and between the two scanning processes on a universal couch-top which also provides landmarks within a coordinate reference arrangement defined in the couch-top. The constraint and landmarking is necessary to ensure that anatomical positioning and features are maintained the same to a first order of accuracy for the CT and MR scans, and for the final radiation treatment. The respective transverse planes of CT and MR images are processed individually by using 'boundary' or 'finger-print' matching techniques in a de-convolution space (DCS) for the CT & MR images. The two sets of de-convoluted maps are then merged together to a second order of accuracy in order that the CT numbers may be transferred over to replace the corresponding MR contrast numbers. Once this has been done, non-transverse (or oblique) planes are obtained from the 2D-MR image or from the re-arrangement of the corresponding voxels of the

3D-MR volume; in the 2D-MR case, a further step of contrast transformation is required.

5 Geometrical distortion in the MR images may be corrected to a certain extent, by a combined spin-echo (SE) and field-echo (FE) scan sequence, with the SE part to correct for the field inhomogeneity in B₀, and the FE part to overcome the impurity or non-linear gradients. The geometrical distortion may also be dealt with
10 empirically using a post-process correction based on a set of phantom data relating to a drum phantom. Movement problems may be suppressed by some scan sequences and/or by gating; for certain parts of the body there is no movement problem for the MR scan.

15 Once the necessary sets of MR images with transferred CT numbers have been obtained, 3D-RTP may be started using the current 2D-RTP software package. The whole '3D-' space is covered through the use of several 2D planes in
20 a pseudo-3D approach.

The results of the combined CT and MR method may be accumulated in a data bank so as to enable the technique to be performed in a so-called stand-alone MR method.
25 For this method the X-tubing coordinate reference arrangement of the couch top is filled with a chemical solution, such as MnCl₂, of appropriate T₁ and T₂ values
to establish a standard in the resulting MR image against which the other components of the image can be related.
30 In this way, it is possible to standardise the contrast numbers of various types of tissue using a standardized 3D-RTP Scan Sequence. A data bank of standardised contrast numbers applicable to various tissues of different patients versus the CT numbers derived from the
35 combined CT and MR method may then be accumulated over a period of time. The data, as 'banded' in the data base, may then be used to enable the stand-alone MR method to

be performed. Interpretation may still be required by an operator or through an 'expert' system to overcome certain ambiguities (for example, in separating bone from air).

5

The matching technique is based on a method using de-convolution space (DCS) in combination with a least-squares curve-fitting method. These methods are separately well known in pure mathematics, but not, as in the present method and system, combined to produce de-convolution maps with the least-squares fitting curve used as a running filter. The filter is used to overcome practical difficulties due to noise within the image pixels.

15

For a mathematical model of the method, consideration is given to a line spread function (LSF) arising in the imaging system from an impulse or δ function; this is represented in Figure 8 of the accompanying drawings.

20

An example of the way in which the LSF of Figure 8 is convolved with an object step-down function to result in a roll-down edge response function (ERF), and how this through the de-convolution process reproduces the LSF, is illustrated in Figure 9 of the accompanying drawings. As illustrated, the LSF is a one dimensional profile across the edge of a contour with its peak position representing the 50%-level point of the ERF.

30

Figure 9 and the mathematical expressions set out below, indicate the way in which the LSF can be recovered from the resulting ERF at a contrast edge. For convolution related to points 1 to n of the step function:

35 at point 1:

$$\sum_0^a L(x) \Delta x = E(x)_1$$

at point 2:

$$\sum_0^a L(x) \Delta x - \sum_{a-1}^a L(x) \Delta x = E(x)_2$$

at point 3:

$$\sum_0^a L(x) \Delta x - \sum_{a-2}^a L(x) \Delta x = E(x)_3$$

5

until at point n:

$$\sum_0^a L(x) \Delta x - \sum_1^a L(x) \Delta x = E(x)_n$$

For de-convolution ie. for the ERF to manifest the LSF:

$$E(x)_1 - E(x)_2 = \sum_{a-1}^a L(x) \Delta x = L(x)_{(a-1)-a} \Delta x$$

10

$$\{E(x)_1 - E(x)_2\} / \Delta x = L(x)_{(a-1)-a}$$

thus:

$$\{E(x)_2 - E(x)_3\} / \Delta x = L(x)_{(a-2)-(a-1)}$$

leading to

$$\{E(x)_{(n-1)} - E(x)_n\} / \Delta x = L(x)_{(n-1)-n}$$

- 15 Thus, the same shape of LSF is recovered by the
de-convolution process independently of the sense of the
ERF; it does not matter whether a roll-up or a roll-down
ERF of a positive or a negative contrast profile, is
involved. This is an important property as a positive
20 contrast contour will have roll-up (from low to high)
ERFs at both ends whereas a negative contrast contour
will have roll-down (from high to low) ERFs at both ends.

In the practical implementation, the LSF is derived from the ERF by de-convolution using a running filter. This enhances the accuracy of the method in overcoming the problem of noise that affects the digitised pixel values of the image. Use is made of a least-squares fitting of a set of data points that advances along the whole length of the function from one point to another.

Assuming that $y = a_0 + a_1x + a_2x^2$ represents the ERF curve a five-point or seven-point fit is used, and the normal equation becomes:

$$\begin{bmatrix} \sum 1 & \sum x_i & \sum x_i^2 \\ \sum x_i & \sum x_i^2 & \sum x_i^3 \\ \sum x_i^2 & \sum x_i^3 & \sum x_i^4 \end{bmatrix} \begin{bmatrix} a_0 \\ a_1 \\ a_2 \end{bmatrix} = \begin{bmatrix} \sum y_i \\ \sum x_i y_i \\ \sum x_i^2 y_i \end{bmatrix}$$

where: i is $1+n, \dots 5+n; \dots$ until $m-5, \dots m$
or
 i is $1+n, \dots 7+n; \dots$ until $m-7, \dots m$
 n is $0, 1, 2, \dots$; and
 $1, \dots m$ is the span of the ERF profile.

The solution may be derived from either:

$$\begin{bmatrix} a_0 \\ a_1 \\ a_2 \end{bmatrix} = \begin{bmatrix} \sum 1 & \sum x & \sum x^2 \\ \sum x & \sum x^2 & \sum x^3 \\ \sum x^2 & \sum x^3 & \sum x^4 \end{bmatrix}^{-1} \begin{bmatrix} \sum y \\ \sum xy \\ \sum x^2 y \end{bmatrix}$$

20

or:

$$\frac{a_0}{\begin{bmatrix} \sum y & \sum x & \sum x^2 \\ \sum xy & \sum x^2 & \sum x^3 \\ \sum x^2 y & \sum x^3 & \sum x^4 \end{bmatrix}} = \frac{a_1}{\begin{bmatrix} \sum 1 & \sum y & \sum x^2 \\ \sum x & \sum xy & \sum x^3 \\ \sum x^2 & \sum x^2 y & \sum x^4 \end{bmatrix}} = \frac{a_2}{\begin{bmatrix} \sum 1 & \sum x & \sum y \\ \sum x & \sum x^2 & \sum xy \\ \sum x^2 & \sum x^3 & \sum x^2 y \end{bmatrix}} = \frac{1}{\begin{bmatrix} \sum 1 & \sum x & \sum x^2 \\ \sum x & \sum x^2 & \sum x^3 \\ \sum x^2 & \sum x^3 & \sum x^4 \end{bmatrix}}$$

For both of these equations to be valid:-

$$\begin{vmatrix} \Sigma 1 & \Sigma x & \Sigma x^2 \\ \Sigma x & \Sigma x^2 & \Sigma x^3 \\ \Sigma x^2 & \Sigma x^3 & \Sigma x^4 \end{vmatrix} \neq 0$$

The gradient at $dy/dx_{(3+n)}$ or $dy/dx_{(4+n)}$ can then be derived and plotted against x for the LSF profile.

5 The graph of dy/dx against x gives the LSF profile. The peak of this profile is located half way between the mid-points of the ascending and descending limbs of the graph and is derived from this as the so-called full-width-half-maximum (FWHM) with an accuracy to the sub-pixel
10 level.

The point spread function (PSF) is the two dimensional profile which may be derived, in practice, from the two corresponding LSFs orthogonal to one another within an
15 image plane. The peak position of the PSF profile is, therefore, from the 'mean' or 'cross-over' of the two peaks or the two LSF profiles. The PSF is obtained in practice from two orthogonal axes in a two-dimensional plane.

20 The waveform relationships between real space (RS), image space (IS), and de-convolution space (DCS) are illustrated at (a), (b) and (c) respectively of Figure 10 of the accompanying drawings. These show the
25 relationship between peak positions p_1 , p_2 , p_1' , p_2' , p_1'' and p_2'' and dip positions d' and d'' in the DCS, with the positions and magnitudes of contrast value of the waveforms in the image space (IS) and real space (RS), and highlight certain useful properties of the process.
30 In particular, the tail-end to tail-end distance of each LSF profile gives a spread of 100% down to 0%, and vice versa, in the corresponding ERF profile in IS. Also, the half way points of the full-width-half-maximum (FWHM) in

the LSF profiles give the peak positions of p_1 , p_2 , p_1' , p_2' , p_1'' , and p_2'' which correspond to the 50% level points of their related ERF profiles; the dip positions d' and d'' in the LSF profiles correspond to the central
5 maxima or minima of the profiles in IS, and the distances of p_1' to p_2' , and p_1'' to p_2'' correspond to the FWHMs of the profiles in the IS and the distances between the edges of the step function in RS.

10 From the above, the generation of the LSF (or PSF) is independent of the roll-up or roll-down nature of ERFs at the edges of the contrast contour. In other words, it is independent of the sense and the absolute value of the contrast numbers within the contour. The peak position
15 of the LSF (or PSF) is the central half-way (50%) point of the roll-up or roll-down ERF which is the optimum position at which the two sets of ERFs are matched together with the overall effect of not distorting the contour shape within the boundary of the ERFs.

20 In order to match or merge together the two corresponding CT and MR images, they are transformed into DCS. In DCS the peaks of the LSFs (or PSFs) mark the central half-way points of the ERFs of the original two-dimensional image
25 planes. These peak positions are then linked together to form the 'de-convolution' or 'finger-print' maps.

Through the assistance of these maps, the two image files may now be matched or merged together to a second order of accuracy. Figures 11 and 12 of the accompanying
30 drawings illustrate one-dimensional and two-dimensional generation of de-convolution or finger-print maps in DCS.

The subsequent transformation of the contrast numbers is guided by a moving sub-set filter conformed by the
35 sub-set mean values and the overall uniformity of the profile of the MR image; this filter is quite similar to the running filter used for the conversion of ERF to LSF.

The number of samples in the moving sub-set filter may also be used for the cut-off rejection of small or 'point' contours, that is to say, spurious contours introduced by noise, within the CT or MR image.

5

A similar technique is employed for MR(Tr) to MR(Oblique) transformation of contrast numbers. Since a high spatial resolution and small partial volume effect are maintained for the thin 2D-MR(Oblique) slice or the 3D-MR(Oblique) plane, the transformation is relatively simple. Through this DCS technique, the high spatial resolution operation is maintained throughout the whole process of re-processing.

15 The combined MR and CT method of the invention requires the matching of corresponding CT and MR images with one another, and it is important that the same 'laid-out' features of anatomy are presented for the each scanning activity, and that the same features are presented again for radiation treatment. The couch-top is used for this and the landmarks and reference axes defined in it are encoded into the images, for use as guides for the repositioning of the patient for any subsequent scanning or other activities, and also for the first order alignment in the initial matching of CT and MR images (allowing more refined matching to the second order of accuracy using the DCS technique, to be initiated).

30 The form of the couch-top used will now be described with reference to Figures 13 to 15 of the accompanying drawings.

Referring to Figure 13 to 15, the couch-top has a flat upper part 100 that is carried by a curved lower part 101 which fits to the scanner table of the CT or MR installation. The patient lies on the flat part 100 and is constrained on it (by an arrangement not shown), so as

to reproduce the same anatomical features for scanning and treatment; the treatment table similarly has a flat surface.

5 An X-tubing array 102 is embedded into the upper part 100. The tubing of the array 102 is of plastics or glass having an external diameter of up to 6 mm and an internal diameter of up to 3 mm, and is filled with a chemical solution of $MnCl_2$. The cross-sectional image of such
10 tubing will show a positive profile for MR and a negative profile for CT. However, the DCS technique of processing these images pin-points their centres irrespective of the contrast values represented.

15 As shown in Figure 14, the patient is constrained by the constraint arrangement to lie lengthwise of the flat part 100 (ie along the z-axis of the coordinate system defined in the couch-top) and centrally of it (ie with respect to the lateral x-axis); the patient's head is placed on a
20 head-rest 103 suitable for the scan, CT or MR, to be undertaken. Three landmarks/constraint levers 104, two under the armpits and one at the crotch between the thighs, are also used to position the patient as well as to provide some form of conformation with respect to the
25 reference coordinate system. Air-compressed cushions (not shown) may be gently pressed in from either sides to conform the limbs and hence the body along its length.

30 The X-tubing array 102 inside the bottom flat part 100 may be used as the reference system for the transverse and sagittal planes, and the oblique planes between them. The flat surface of the part 100 or the flat contour of the patient's skin surface in contact with it may be used as the datum for the coronal planes. However, an extra
35 small X-tubing array (not shown) may be placed above the patient using the three landmarks as the supporting legs; this enables the small-angled oblique planes

between the coronal and transverse planes to be monitored.

5 If the internal-bore diameter of the magnet of the MR scanner is large enough, an extra X-tubing array 105 (Figure 15) may be used along either side of the patient to help in monitoring the small-angled oblique planes between the coronal and sagittal planes. Furthermore, a curved X-tubing array (not shown) may be placed inside 10 and on the bottom of the head rest (or head coil) 103 used for the MR scan for the purpose of clearly establishing the coordinates of the head.

15 The material of the couch-top is carbon fibre so as to ensure a low linear attenuation coefficient with transparency to X-rays. Edges of the couch-top are smoothly rounded to avoid sharp edges that would cause streaking in the CT image. Non-magnetic material is used throughout to avoid disturbing the uniform magnetic 20 field, B_0 , used for MR.

The radiation treatment table needs to be capable of both translational, "in and out" movement as well as rotational movement to cover the 3D planning and 25 treatment.

A final empirical correction of any residual geometrical distortion may be carried out using a cylindrical or elliptical drum phantom. A cylindrical drum phantom is 30 illustrated in Figures 16 and 17, and an elliptical drum phantom in Figures 18 and 19 of the accompanying drawings.

35 Referring to Figures 16 and 17, the cylindrical drum phantom, which may be of about 160 mm in length and is for use with the head coil of the scanner, includes a central rod 200 of polymethyl methacrylate (as sold under

the Registered Trade Mark PERSPEX) having a diameter of about 2 mm and defining the phantom centre, Cph. The body of the phantom is built up as concentric sections about the rod 200, each comprising evenly-spaced blocks 201 of polymethyl methacrylate and with the blocks 201 of each section in exact register radially with the gaps 202 between the blocks 201 of the preceding section. The central region 203, having a diameter of about 20 mm, may be left blank (since the magnetic field within the central region is uniform to within 0.1 ppm, so no B0 correction is required). The gaps 202 and the central region 203 are filled with an NMR solution (e.g. MnCl_2 or CuSO_4).

Peak positions of the PSFs in the DCS map for the drum are derived and stored. This data is then corrected for each point of the map with respect to the iso-centre of the scanner system, Ciso, using the following vector equation:-

20

$$\Delta r = r_2 - r_1$$

where Δr is the correction vector between Ciso and Cph; and r_2 and r_1 are the vectors to the relevant point of the map from Ciso and Cph respectively.

The elliptical drum-phantom of Figures 18 and 19 is constructed in essentially the same manner as the cylindrical drum phantom, and is for use with the body coil of the scanner. The major and minor axes are chosen according to the field of view being studied.

As an alternative, the cylindrical and elliptical drum phantoms may be constructed as shown in Figures 20 and 21, using thin plates 300 of polymethyl methacrylate instead of the blocks used for the phantoms of Figures 16

to 19. The plates 300, which may have a thickness of between 1.5 mm and 2.0 mm, are spaced radially from one another by gaps 301 that, together with the central region 302 are filled with appropriate NMR solution.

5

CT scan images are not subject to geometrical distortion so CT scan data accumulated using the phantoms, and the DCS profile data consequently derived and stored, are used as a datum or reference set for correction of the MR scans. Such corrections also automatically correct for manufacturing tolerances, and the DCS data derived from the phantom through the CT and MR scans represents a reference situation devoid of motion.

15

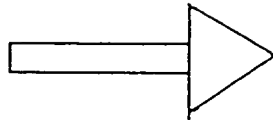
The DCS technique described above is applicable further in diagnosis and treatment. In particular, it may be used to assist visual detection of gradual change of contrast level in scan images. For example, as illustrated in Figure 22 of the accompanying drawings, the 100% point at which the roll-off of a lesion profile begins can be readily plotted from the DCS profile, together with the 50%-level point, and then the 0% point where the profile has completely levelled off. Graphic representations of these various points can then be plotted out to facilitate diagnosis, as illustrated in Figure 23 of the accompanying drawings, in the form of a string of characters 400 delineating the 100%-level, a string of characters 401 delineating the 50%-level and a string of characters 402 delineating the 0% level.

30

In current radiation treatment planning, a composite volume leaving a margin of 15 mm around a lesion, is used. To be effective, the tail-end of the lesion growth is to be within the treatment-beam path, and in this regard there is advantage in extending the DCS-profile plot of Figure 22 to include a point which is a distance (d) beyond the 0%-level. This is then used for diagnosis

35

and treatment, to establish an outer string of characters 403 in the graphical representation of Figure 23. The distance (d) covers the penumbra-spread as well as any possible mis-alignment by the operator.

Object DomainDelta
Function

by Image Transfer:

(The transfer loss is caused by the limitation factors such as the finite size of the energy source, detector, sampling frequency, display density and software filter function etc ..)

by the Convolution of
the PSF with the ERF

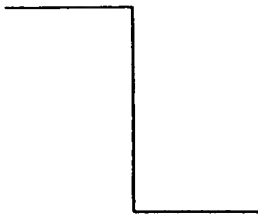
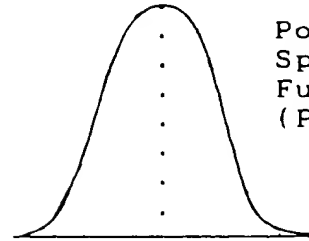
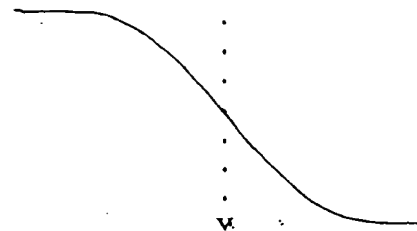
Edge Response Function (ERF)
(in Object Domain)

Image Transfer:

via the Spatial
Resolution Enhancement
Process (i.e. by the
re-arrangement of
the digital data)

Image DomainPoint
Spread
Function
(PSF)

ERF (in Image Domain)



PSF in (solid line)
is recovered by the
De-convolution Space
(DCS) Technique
and the image ERF
(in dotted line)
is super-imposed
into the graph.

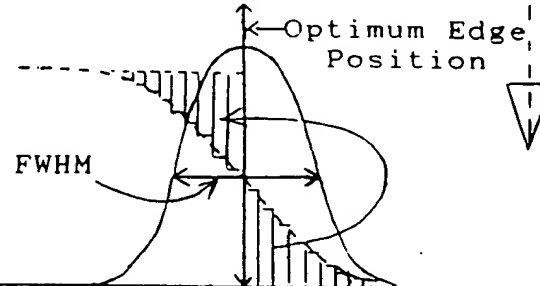


Fig. 1

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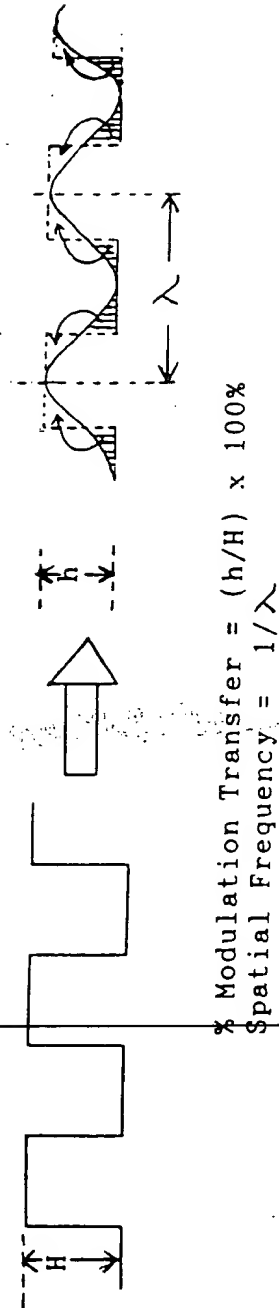
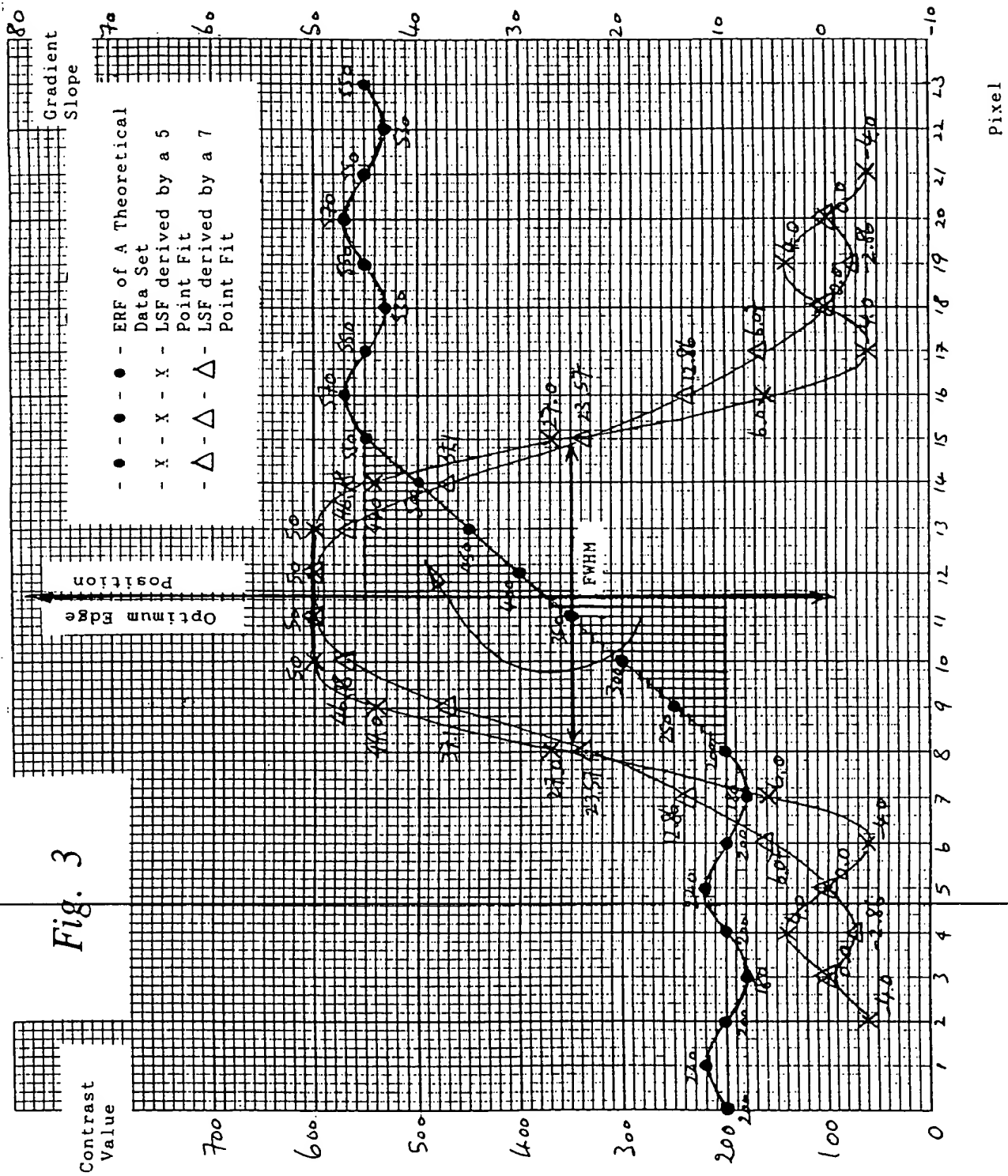


Fig. 2

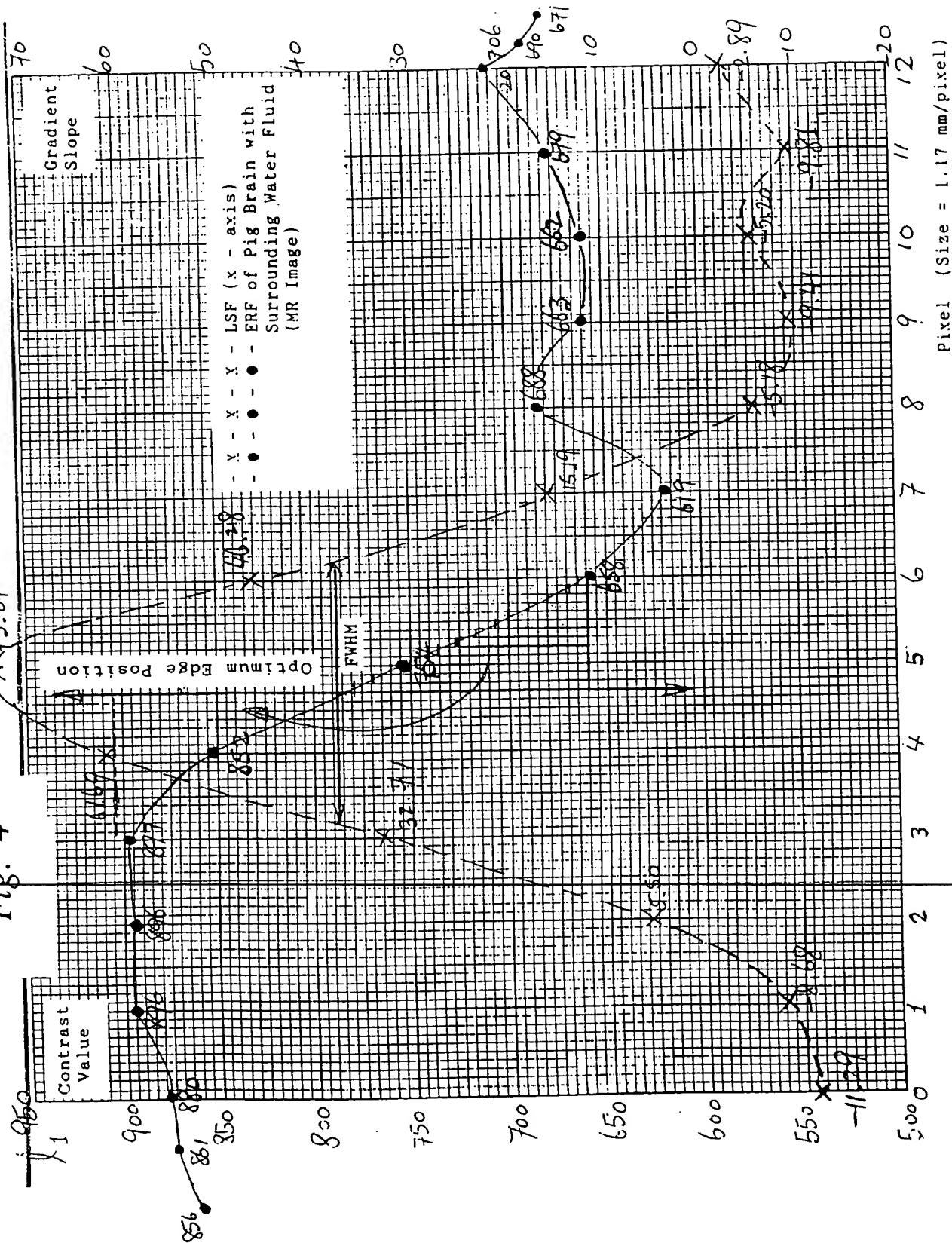
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Fig. 4

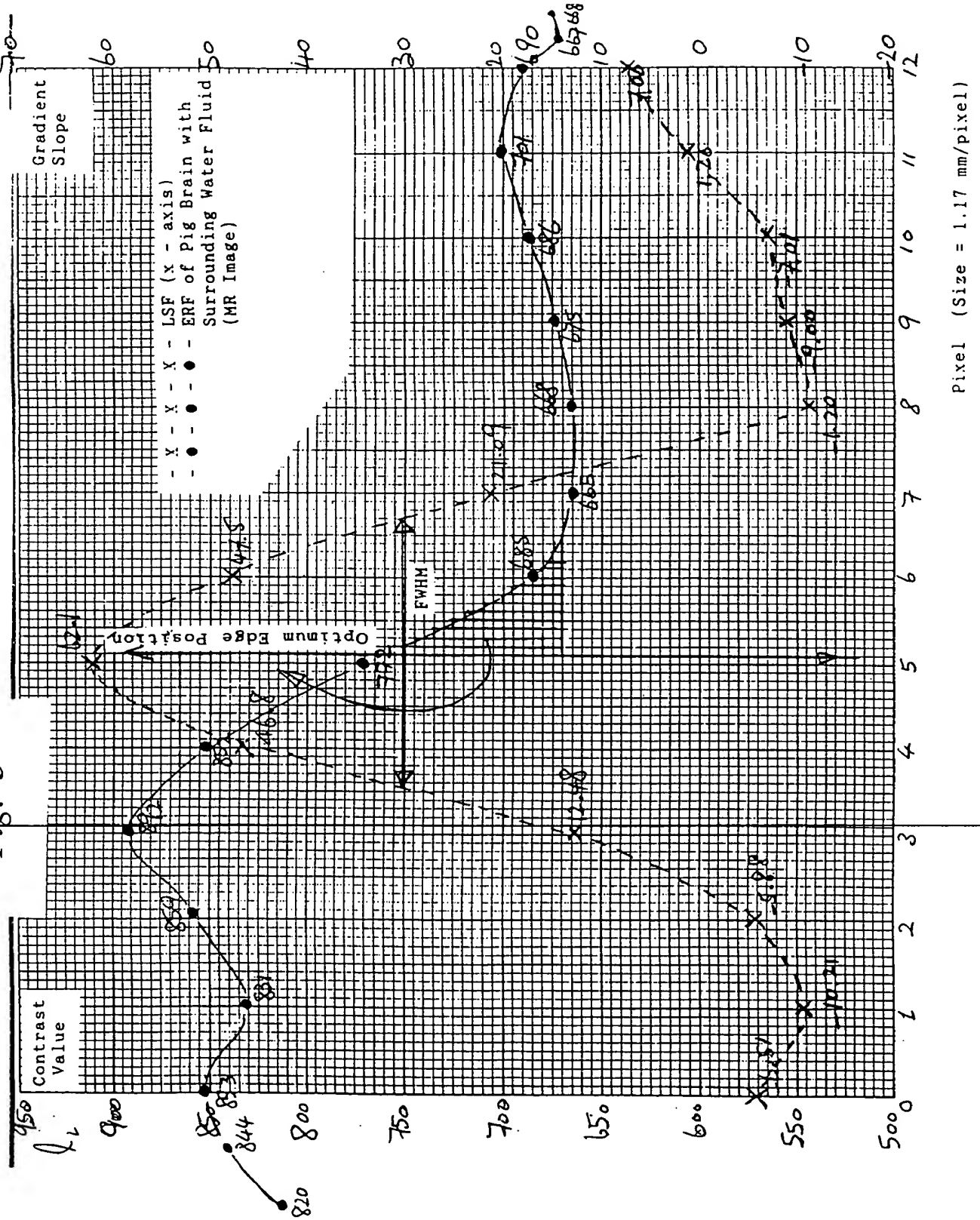


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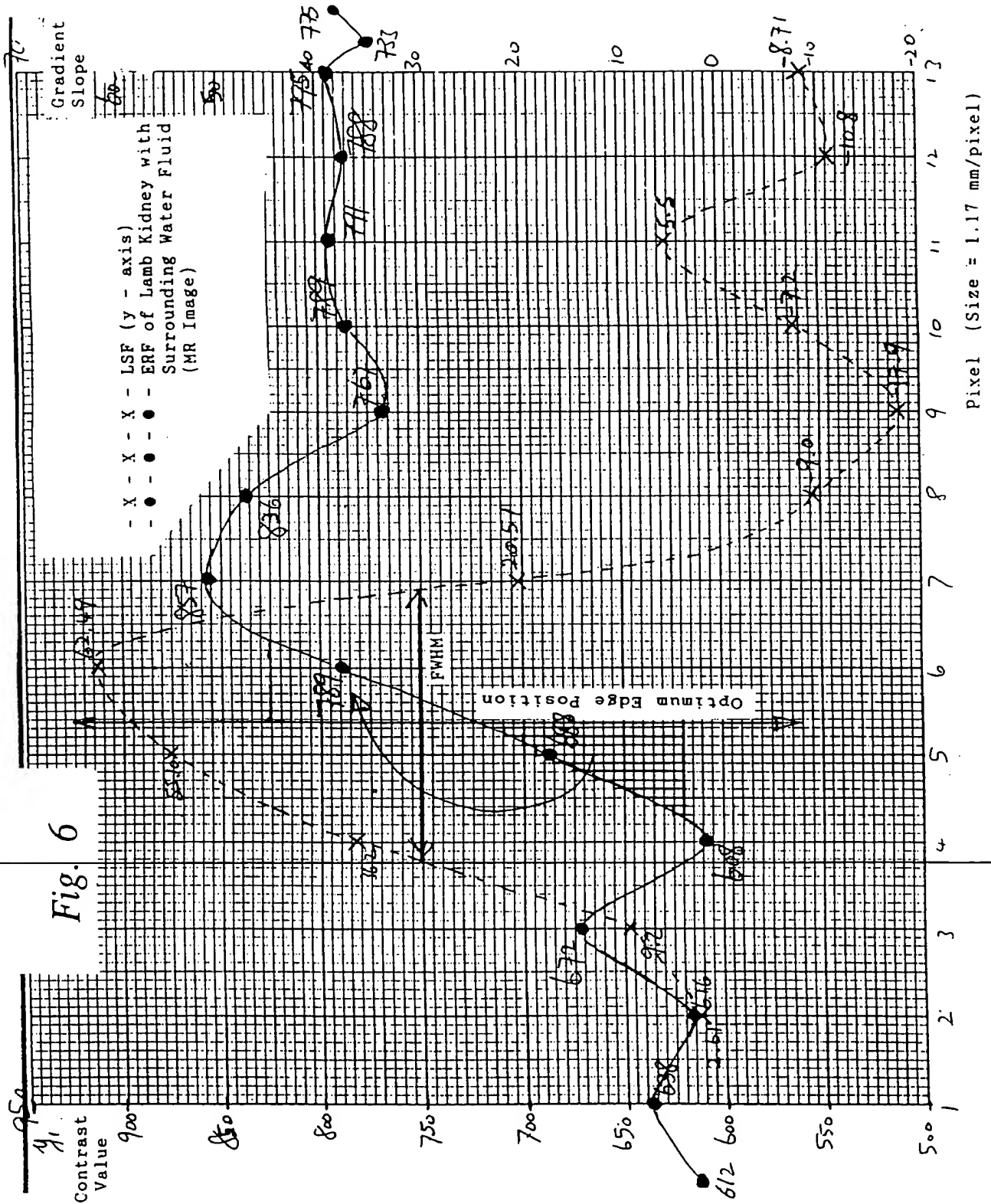
Fig. 5



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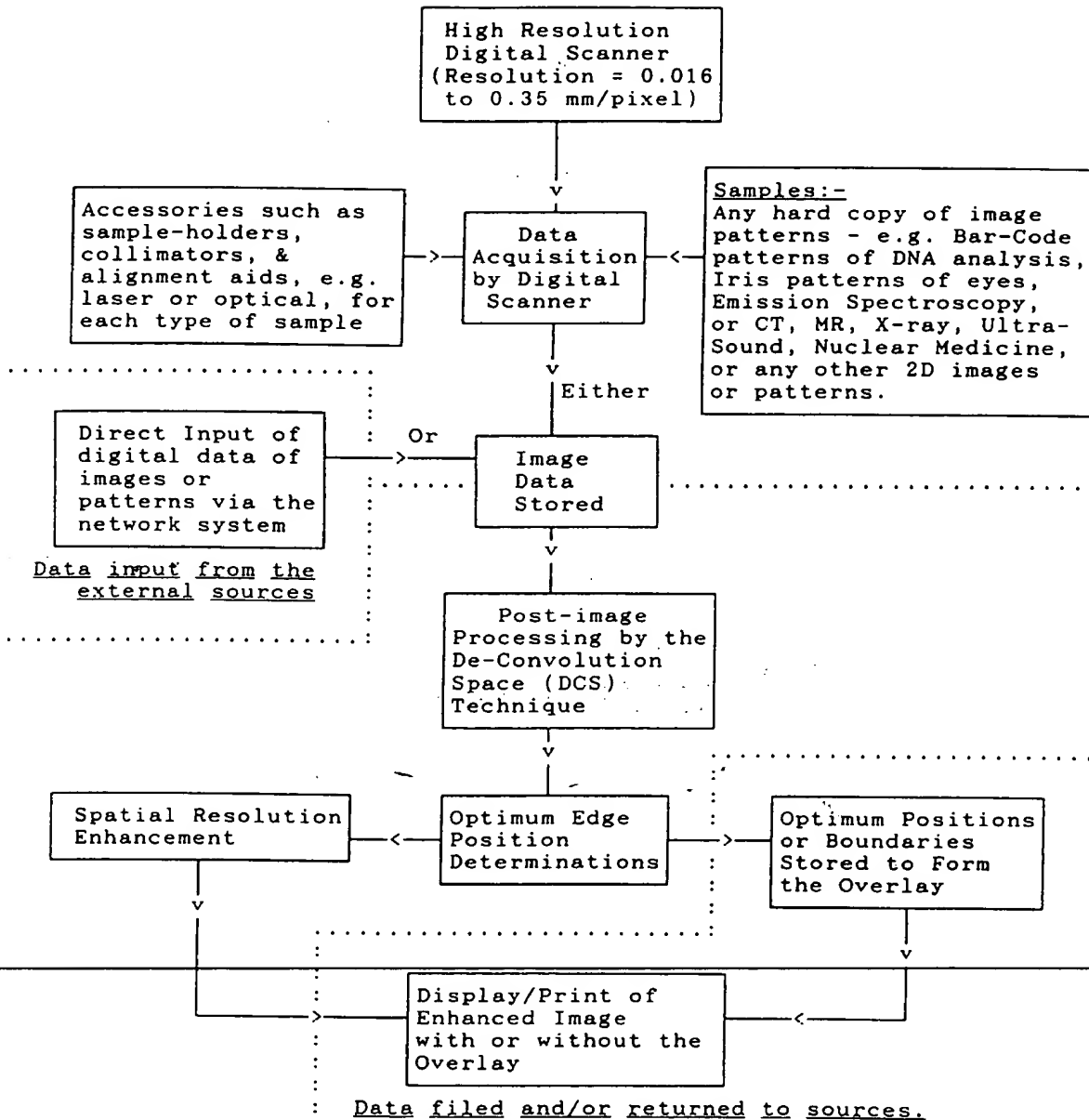


Fig. 7

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Image Domain

TRANSFER



LSF

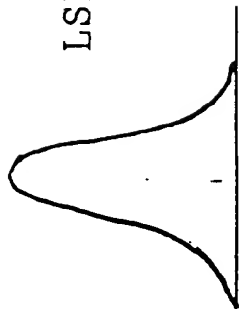
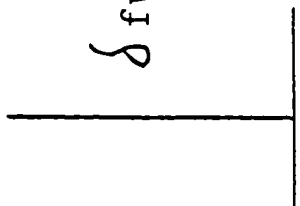


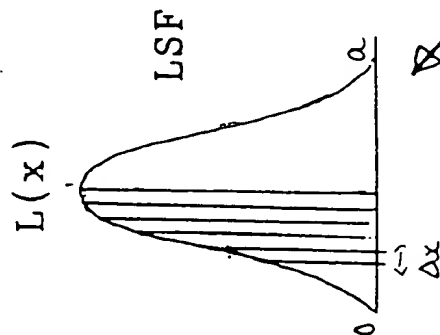
Fig. 8

Object Domain

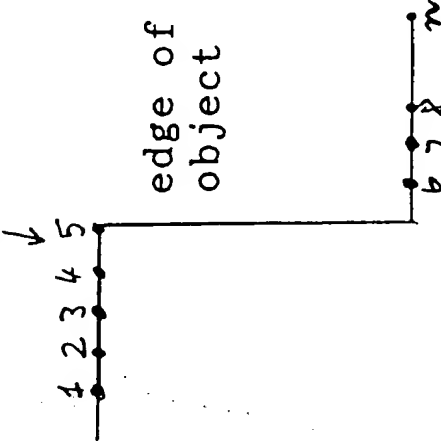
δ function



convolve with

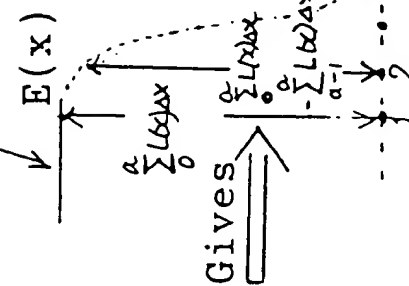


Step-down STEP function



edge of object

Roll-down ERF



De-Convolution (Differentiation)

Fig. 9

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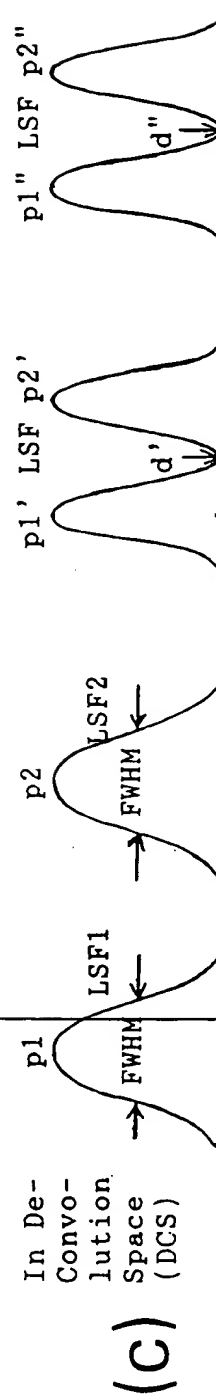
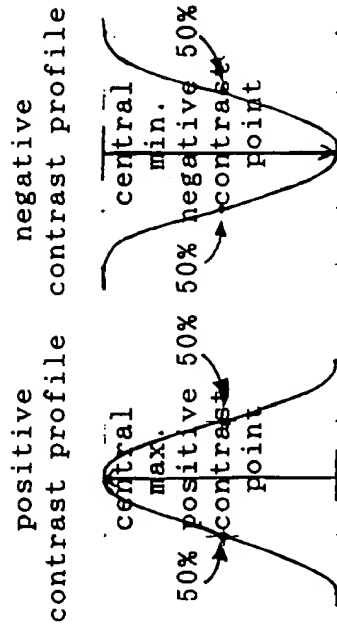
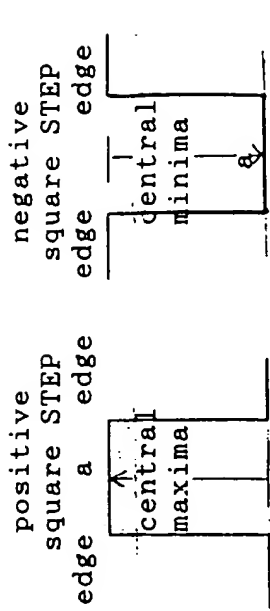


Fig. 10

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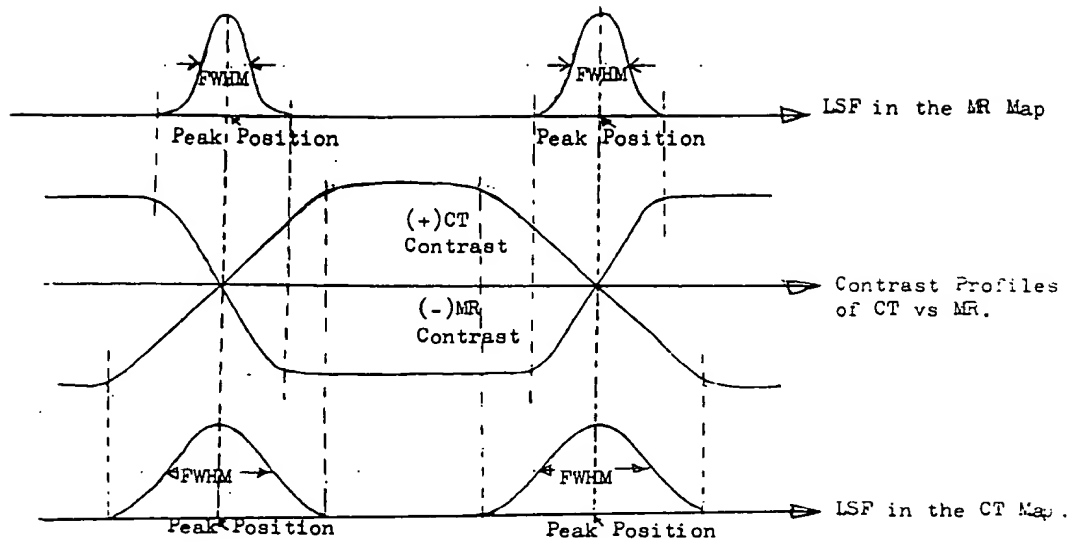


Fig. 11

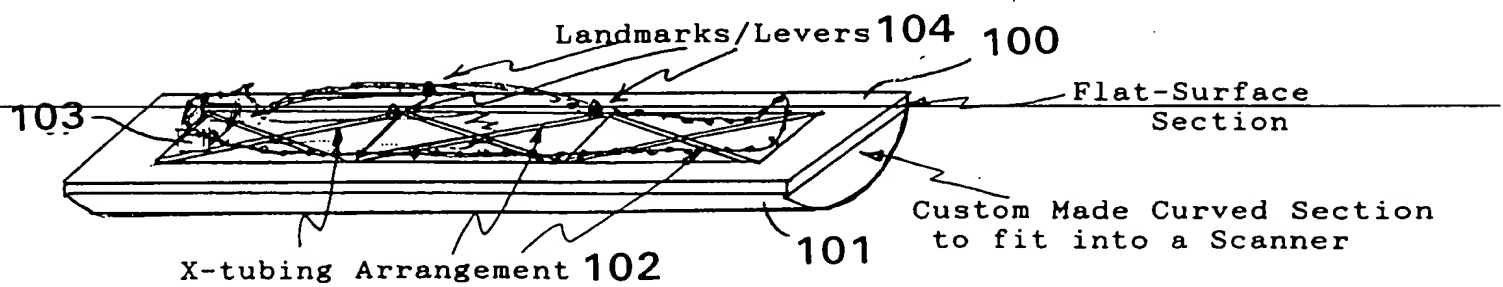
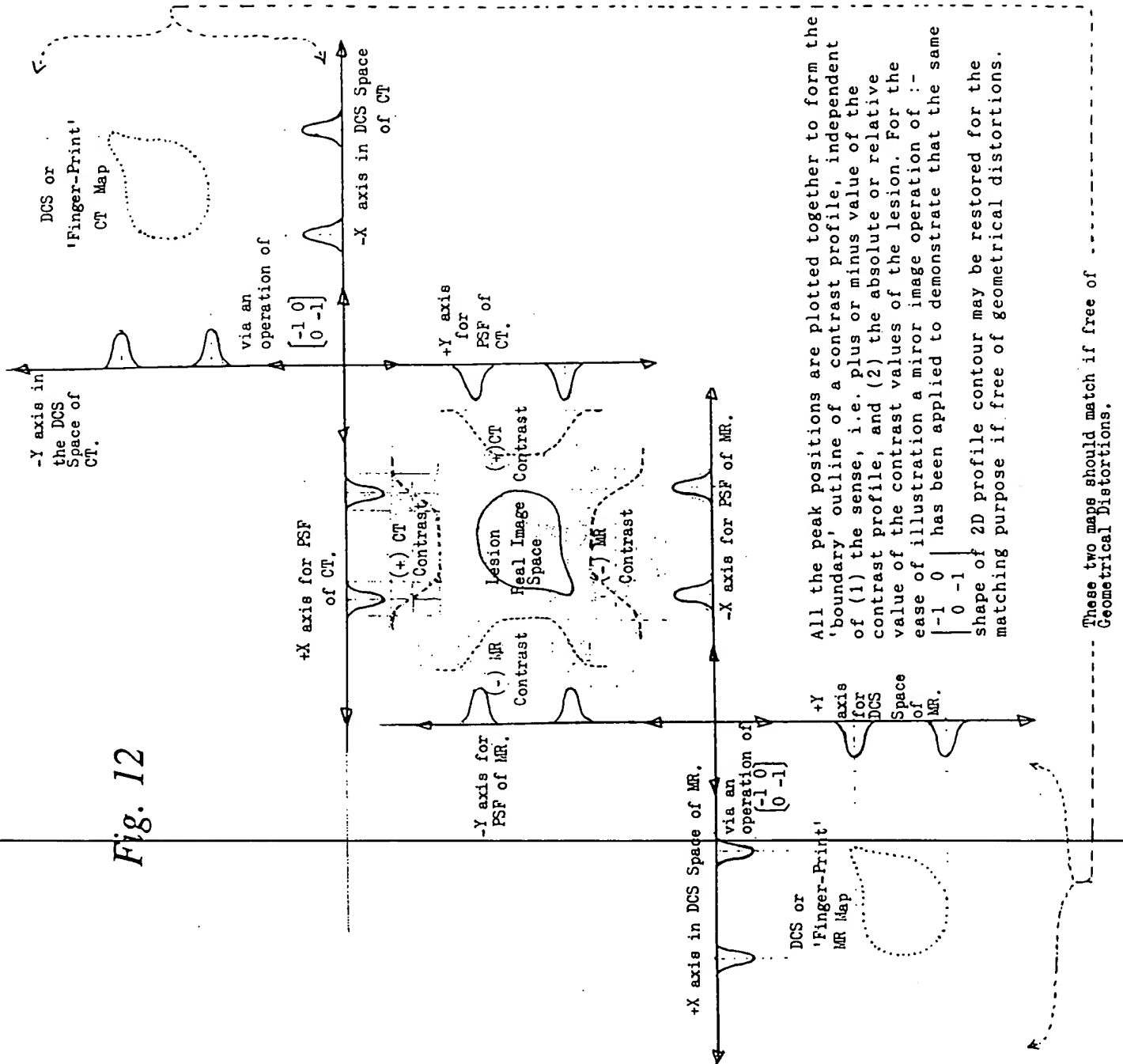


Fig. 13

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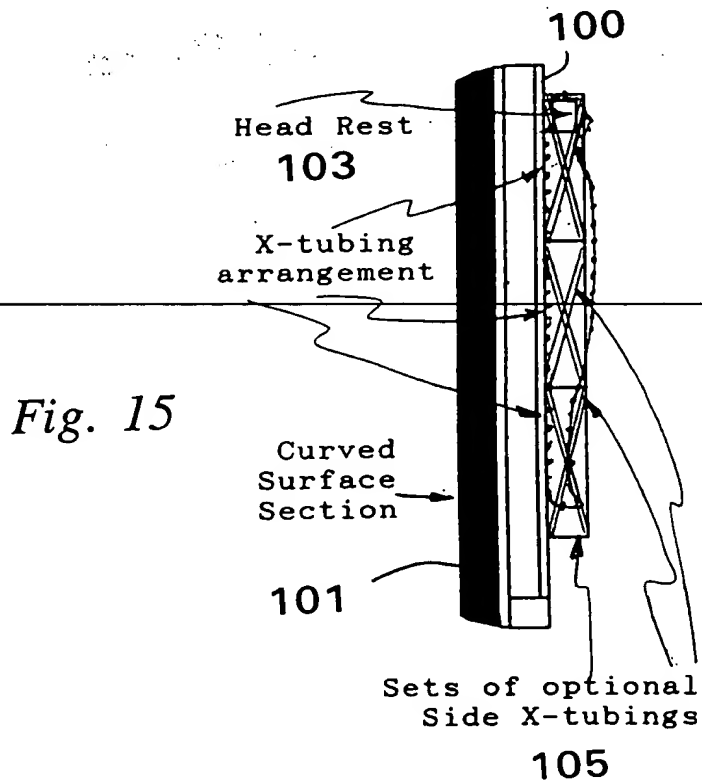
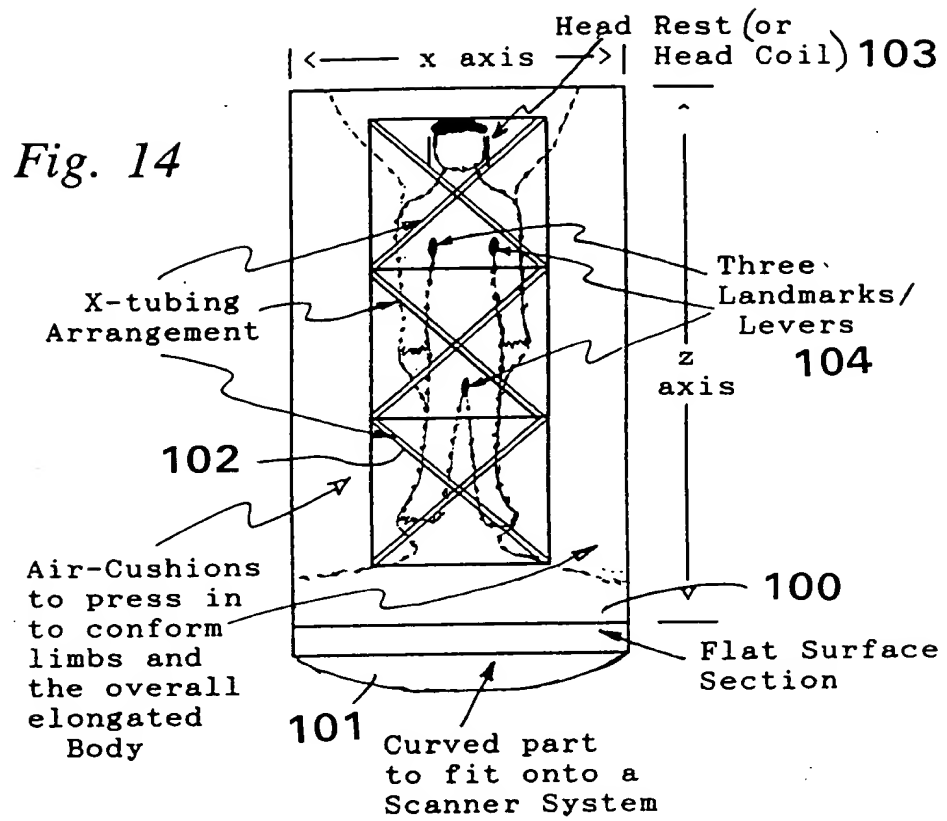
Fig. 12



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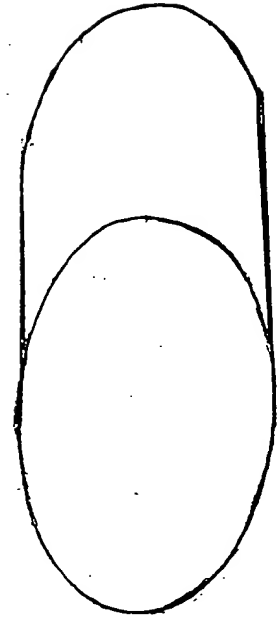


Fig. 16

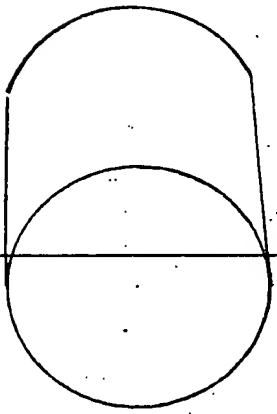


Fig. 18

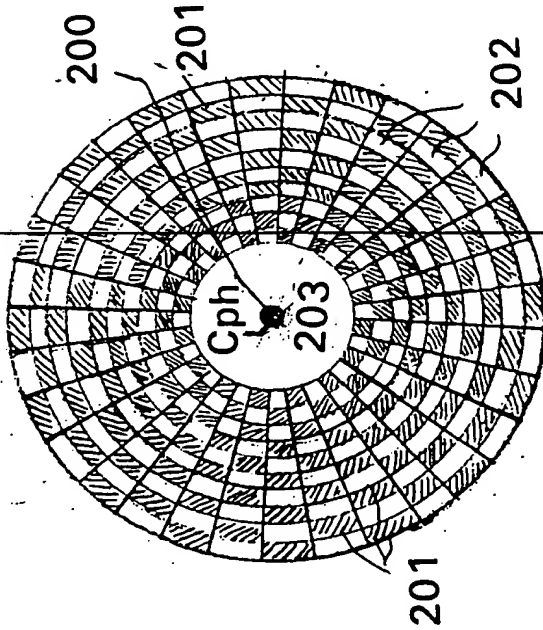


Fig. 17

Fig. 19

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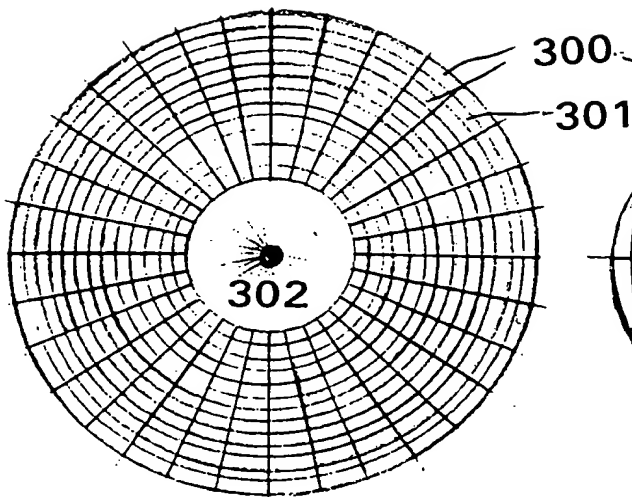


Fig. 20

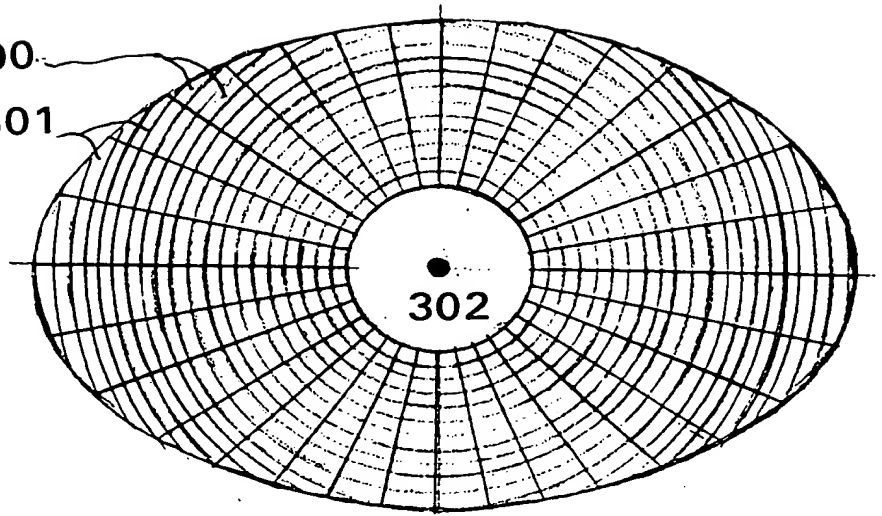


Fig. 21

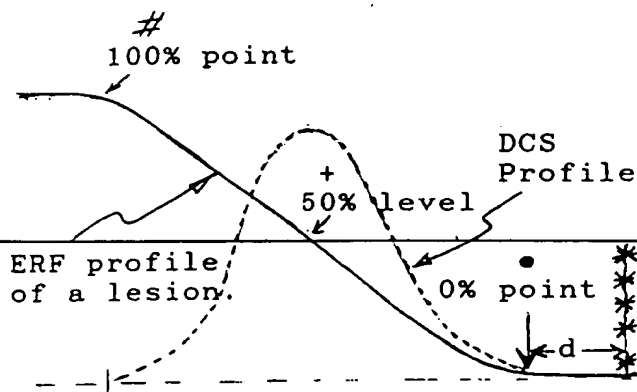


Fig. 22

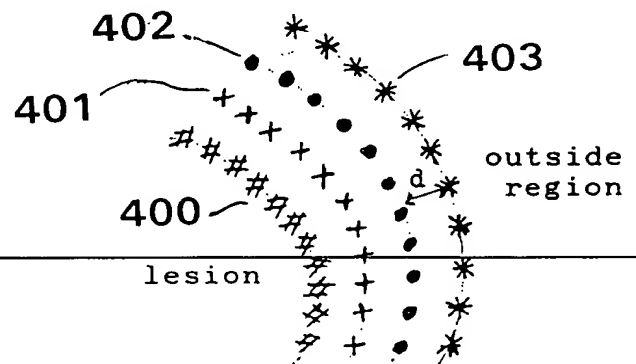


Fig. 23

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